Introduction of sample interval modulation for the simultaneous acquisition of 3D displacement data in MR Elastography

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Introduction: In Magnetic Resonance Elastography (MRE) [1] external vibrations are introduced into the tissue under examination. The tissue vibrations are encoded in the MR signal phase φ using standard MRI sequences upgraded with motion encoding gradients (MEG). Hence, tissue mechanical parameters can be calculated from the acquired wave fields. The analysis of MRE data with one motion-encoding direction has revealed the correlation of pathophysiological changes and the mechanical behavior of diverse organs [2]. However, more and more the 3D displacement field is acquired to separate the operator [3] and to assess tissue compressibility [4]. Problem: In MRE, measurement time is critical. Besides cost factors, long acquisition times potentially decrease the measurement accuracy, since motion may occur and cause misalignment of the images. Further, in conventional 3D MRE, the components of the tissue displacement are acquired in three individual temporally-resolved MRE experiments. Therefore, the components, although attributed to the same point in time, were actually acquired in different physiological states. Objective: We developed a concept for the MEG arrangement that is capable of encoding three spatial components of a monofrequency tissue vibration simultaneously. We name our approach Sampling Interval Modulation (SLIM)-MRE, since the individual displacement components are observed using different time discretization intervals. In doing so, the components are modulated with different frequencies in the MR signal phase φ expressed as a harmonic function of the start time of the MEG. Thus, all displacement components are acquired faster than in conventional MRE and can be derived from the same temporally-resolved MR phase images. We present for the first time, to our knowledge, 3D displacement data that were acquired simultaneously and stored in the same k-space.

Theory: Below, the index j=1, 2, 3 corresponds to the x-, y- and z-direction in the Cartesian system, and to the read-, phase- and slice-direction in the scanner system, respectively. The basic equation of MRE is represented by eq. 1 [1], which describes the encoding of the displacement u_j of an isochromat in the MR signal phase φ by applying a magnetic field gradient G_j. Herein, the gyromagnetic ratio of the proton, the duration and the start time of the MEG-component in the j-direction are denoted by γ, T and s_j, respectively. Of special note, in SLIM-MRE, T is kept constant for all MEG-components, while s_j may vary, and we assume sinusoidal functions of the same frequency f for the vibration and for the three MEG-components. Eq. 2 represents the solution of eq. 1 and comprises the initial mechanical phase φ_0, the amplitude u_0 and the encoding efficiency ε of the displacement component u_j. Temporal resolution is expressed by a variation of the MEG-start time. In SLIM-MRE, the sampling interval of the MEG-start time is different for the three components. We set for the sampling interval Δt = f/T, j = 1, 2, 3, with N being the number of samples and s_0 = nΔt, n = 0, 1, ..., N-1. It is directly perceptible from eq. 2, that the three displacement components are encoded with different “apparent frequencies”, specifically with the 1st, 2nd and 3rd harmonic. The individual components can thus be decomposed by applying a discrete Fourier transform to φ. An example for the arrangement of the MEGs in SLIM-MRE is shown in fig. 1.

Methods: The experimental setup for a similar MRE experiment in the same 11.7 T Bruker vertical MRI system has been described before [5]. We prepared an inhomogeneous phantom consisting of agarose beads (0.7% by volume) as an inhomogeneous phantom consisting of agarose beads (0.7% by volume) for SLM-MRE and can be derived from the 3D displacement field that is acquired to separate the operator [3] and to assess tissue compressibility [4]. Problem: In MRE, measurement time is critical. Besides cost factors, long acquisition times potentially decrease the measurement accuracy, since motion may occur and cause misalignment of the images. Further, in conventional 3D MRE, the components of the tissue displacement are acquired in three individual temporally-resolved MRE experiments. Therefore, the components, although attributed to the same point in time, were actually acquired in different physiological states. Objective: We developed a concept for the MEG arrangement that is capable of encoding three spatial components of a monofrequency tissue vibration simultaneously. We name our approach Sampling Interval Modulation (SLIM)-MRE, since the individual displacement components are observed using different time discretization intervals. In doing so, the components are modulated with different frequencies in the MR signal phase φ expressed as a harmonic function of the start time of the MEG. Thus, all displacement components are acquired faster than in conventional MRE and can be derived from the same temporally-resolved MR phase images. We present for the first time, to our knowledge, 3D displacement data that were acquired simultaneously and stored in the same k-space.

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Fig. 1: MEG arrangement relative to the mechanical actuation signal in SLIM-MRE. For better visualization, only one MEG-cycle is illustrated per component and time step s. The start of all MEGs in the first time step (s=0) coincides with n=0. Using f=5kHz, the sampling interval with respect to the start time of the x-, y- and z- MEG corresponds to 25 μs, 50 μs and 75 μs, respectively.

Fig. 2: Complex wave images in the axial plane at 5 kHz acquired using conventional MRE (top) and using SLIM-MRE (bottom). One displacement component is shown per column as indicated in the heading. The ROI is demarcated with dashed lines.